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TITLE: A MULTI-MODE X-RAY IMAGING SYSTEM WITH ONE-SIDED CT CAPABILITY

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APPENDIX B
DOD No. 88.1

U.S. DEPARTMENT OF DEFENSE

SMALL BUSINESS INNOVATION RESEARCH (SBIR) PROGRAM
PHASE 1 - FY 1988
PROJECT SUMMARY

Topic No. 192 Military Department/Agency ARMY.

Name and Address of Proposing Small Business Firm

Rayex Corporation
9701 Fields Road #1605
Gaithersburg, MD 20878

Name and Title of Principal Investigator

Dr. J. W. Motz, Research Director

Proposal Title

A PORTABLE MULTI-MODE X-RAY IMAGING SYSTEM WITH ONE-SIDED CT CAPABILITY

Technical Abstract (Limit your abstract to 200 words with no classified or proprietary information/data.)

The goal of this project is to develop an X-ray imaging system with three operating modes, including (1) a CT Mode with an open one-sided geometry providing complete or partial CT type image slices with arbitrary orientations and shapes, (2) a Transmission Mode providing conventional transmission images, and (3) a Fluoroscopy Mode providing real-time images. Because this versatile, open system offers immediate access to wounded soldiers with one third the weight and cost of conventional CT systems, it is especially suitable for combat casualty care.

This Phase I program was carried out with a simplified one-sided CT system which can provide tomographic images with X-rays backscattered from different body volume elements penetrated by the X-ray beam. The X-ray image sensor consists of an array of low efficiency plastic scintillation detectors which select and count the scattered X-rays providing the image signal. With spatial resolution indicators used as test objects, the X-ray exposures and the digital outputs of the X-ray detectors were measured to determine the signal-to-noise ratios, the spatial resolutions, and the minimum required X-ray exposures. The results show that with the use of high efficiency detectors, this one-sided CT system is capable of obtaining spatial resolutions from 2 to 5 Line Pairs per cm with an average patient dose per slice of less than 2 rads per image slice, or a spatial resolution greater than 7 Line Pairs per cm with an average patient dose of 5 rads per slice, which is the dose requirements for present CT systems. These results establish the technical feasibility and potential advantages of this one-sided system for medical diagnostic services.

Anticipated Benefits/Potential Commercial Applications of the Research or Development

The development of a radiological diagnostic imaging system that is rugged, portable, easily accessible, and versatile enough to provide both CT type images for the detection of internal bleeding and nonmetallic foreign bodies and conventional transmission images, will greatly enhance the diagnoses, triage, and surgical care of combat casualties at Medical Clearing Stations and Mash Hospitals. In addition, the low cost, low mass, and cube of the system, and its ability to provide digitized images, will make it an important diagnostic tool in the equipment inventory of the Emergency Rooms in civilian hospitals.

List a maximum of 8 Key Words that describe the Project.

Tomography, Backscattered X-rays, Multi-mode, One-sided CT, X-Ray Slot Camera, Digital Radiography

20 February 1990 Phase I Final (8/15/88 - 2/15/89)

A Multi-Mode X-Ray Imaging System with One-Sided CT Capability

Contract No. DAMD17-88-C-8203

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Tomography, Backscattered X-rays, Multi-mode, One-sided CT, X-ray Slot Camera, Digital Radiography, RAD II, SBIR

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I. INTRODUCTION

The development of a radiological diagnostic imaging system that is rugged, light-weight, and low-cube, and that has a digital format with the capability of producing both conventional and CT-type images will greatly benefit the military services. Such an imaging system could be freely moved to field stations and could provide the most advanced radiological diagnostic services even at the smallest medical facilities. With tomographic capability, this system will be able to readily detect internal bleeding, embedded nonmetallic foreign bodies, and damage to internal organs. For the physicians and surgeons at the Medical Clearing Stations and the Mash Hospitals, this capability will greatly enhance the diagnosis, triage and surgical care of combat casualties. As a consequence, there will be a reduction in the mortality rate and an increase in the "return to duty" rate for wounded military personnel.

The Rayex Corporation proposes to develop a versatile x-ray imaging system which is specially designed for combat casualty care. This system, as illustrated in Figures 1 through 6, will have the following features:

* ***Three distinct diagnostic operating modes***
  (1) The CT Mode with an open one-sided geometry providing complete or partial CT-type image slices with arbitrary orientations and shapes as shown in Figures 1, 2, 3, 4, and 5.
  (2) The Transmission Mode providing conventional transmission images with either screen-film cassettes or digital image sensors as shown in Figures 2 and 6.
  (3) The Fluoroscopy Mode providing real-time transmission images as shown in Figure 6.

  All of the above capabilities which are presently provided by different x-ray systems are combined into the single Rayex Multi-mode X-ray Imaging System. In addition, the capability of taking partial slices will permit the radiological examination to be confined only to the critical body regions of interest. This feature of the Rayex System greatly reduces the total body exposure received by the patient in comparison to the exposure received with present day CT systems which require complete axial slices over a large section of the body. No x-ray imaging system can presently offer this unique capability.

* ***Highly Portable***
  The total weight of the Rayex System *(exclusive of the power supply)* will be approximately 400 pounds or less than one third the weight of present CT scanners.

* ***Highly Accessible***
  The open one-sided operation of the Rayex System will provide immediate and easy access to wounded soldiers and will greatly expedite the process of diagnosis, triage, and treatment.

* ***Inexpensive***
  The total cost of the Rayex System will be less than one-third the cost of current CT scanners.

* ***Reduced Patient Time***
  The Rayex System will be able to throughput patients faster then present CT systems, partly because of its one-sided easy accessibility *(see Figure 1)*. In addition, the capability of the Rayex Imaging System to do direct partial CT-type tomographic slices of only critical regions will drastically reduce the time required for many patients who do not require full axial slices for proper diagnosis. *(Present CT systems can only take full axial slices whether they are actually needed or not.)*

* ***Variable Spatial Resolution for CT Images***
  The Rayex System will have the flexibility of changing the spatial resolution of CT images from 2 to 5 Line Pairs per cm by simple changes in the apertures of the slot and the x-ray beam collimator. The low resolutions *(such as 2 LP/cm)* permit fast scans with low exposures. The high resolutions *(such as 5 LP/cm)* require slower scans and higher exposures. For partial body scans, the reduction in the irradiated body volume will permit higher x-ray exposures with less risk and consequently higher spatial resolutions *(up to 5 LP/cm)*.

* ***Fast Scan for Rapid Emergency Diagnosis***
  For rapid emergency diagnosis of wounded soldiers, the Rayex System will provide fast scans for quick detection of internal bleeding. Rapid diagnosis can be achieved by utilizing the one-sided accessibility of the system and by using only partial body scans with low resolution.

* ***Slow Scan for Fine Anatomic Detail***
  The Rayex System will have the capability to do direct partial CT-type tomographic slices which can be confined only to regions of interest. With a slow scan and higher exposures, the Rayex System will be able to produce high resolution images of detailed anatomic structures *(5 LP/cm or greater)*. These images will be superior to those produced by current CT systems and will greatly enhance the diagnosis, triage, and treatment of patients with wounds in areas that have fine, complex anatomy such as the eye, the brain, the spinal cord, and the ear.

The multiple operational modes, of this proposed "all-in-one" field diagnostic imaging system is made possible by combining proven technologies with a new technology. The new technology involves collection of backscattered radiation *(see Figures 3 and 4)* to generate information which is then processed to yield the desired tomographic images. The tomographic capability *(full or partial)* of this unique imaging system will also make it possible to readily detect bleeding tracts and embedded nonmetallic foreign bodies such as glass or plastic. The Rayex System will also produce fluoroscopic images, or conventional transmission images, either on a film or digitized format. This unique blend of existing and new technologies enables the Rayex Multi-mode, CT Imaging System to operate in the three separate and distinct operational modes. Furthermore, in contrast to the conventional CT technique, the use of backscattered radiation makes one-sided operation possible when doing complete or partial CT-type tomographic slicing as there is no need to have an x-ray sensor surrounding the patient. Thus, this new technology will greatly reduce the weight and cube of the system, and will greatly facilitate its use with seriously injured patients who are on a field litter, gurney, or operating room table since these patients will not have to be lifted off the field litter and placed in the aperture of a conventional CT imaging system. Finally, all image information can be generated in digitized form and will be directly available for digital information manipulation, networking, and storage.

The Rayex Multi-mode X-ray Imaging System will provide a new one-sided CT technology, which is based on the formation of tomographic images with x-rays that are backscattered *(or reflected)* from the different body volume elements penetrated by the x-ray beam *(see Figure 3 and 4)*. Previous attempts to produce images with backscattered x-rays have not been practical because all of the different methods had inherently low sensitivities and consequently required excessive x-ray exposures and exposure times. In comparison, the Rayex One-sided CT System, which is based on patents issued to Dr. M. Danos, has sensitivities that are four to five orders of magnitude greater than any previous backscatter imaging system. This greatly increased sensitivity will allow the Rayex System to operate with exposures that are comparable to present-day CT systems for the same image quality.

II. PHASE I TECHNICAL OBJECTIVES

The Phase I Program is intended to demonstrate the technical feasibility and the potential advantages of the Rayex patented one-sided CT system over conventional CT systems for combat casualty care. This program will be carried out with a simplified version of the Rayex CT camera *(Slot Camera)* such that the imaging system will consist of a limited array of 20 plastic scintillation detectors which will be used as the X-ray pixel detectors for the digital solid-state imaging panel. Although these scintillators have a low X-ray detection efficiency, they are cheap and readily available, and therefore can be utilized within the time and cost limits imposed in the Phase I program. It is anticipated that in a Phase II program the array of 20 plastic X-Ray detectors will be replaced with an array of approximately 100 high efficiency X-Ray detectors.

The technical objectives of the Phase I program are as follows:

a. Determine the patient X-ray exposures required for obtaining spatial resolutions in the region from 2 to 5 Line Pairs per cm with test objects containing spatial resolution indicators.

b. Determine the final patient exposures required for obtaining spatial resolutions in the region from 2 to 5 Line Pairs per cm with specific high efficiency X-Ray pixel detectors utilizing well-known conversion factors.

c. Determine scan times required for tomographic image slices corresponding to spatial resolutions in the region from 2 to 5 Line Pairs per cm.

d. Design a high detection efficiency imaging panel which can provide complete body-slice images and spatial resolutions in the region from 2 to 5 line pairs per cm for the Phase II Program.

III. PHASE I PROGRAM

The Phase I Program was carried out by completing the following tasks:

a. Adapt a breadboard prototype of the one-sided CT system designated here as the Slot Camera system.
b. Design and construct test objects to demonstrate the spatial resolution capabilities of the Slot Camera system.
c. Measure X-ray exposure rates provided by the X-ray source for different filtrations and kilovoltages.
d. Determine the detection efficiency of the X-ray detector elements used in the digital imaging panel of the Slot Camera.

e. Measure the digital outputs of the X-ray detector elements and the corresponding X-ray exposure rates for spatial resolutions of 2, 3.3, and 5 Line Pairs per cm.
f. Determine the minimum X-ray exposures and scan times required for different spatial resolutions.

A description of the preceding tasks and the pertinent experimental details is given in the following sections.

A. THE SLOT CAMERA

The basic experimental arrangement of the Slot Camera System is shown in Figures 7, 8, and 9. In this system, the collimated X-ray beam from the X-ray source penetrates through the object of interest, and the X-rays scattered at approximately 90 degrees[1], pass through the slot aperture to the detector array. Each detector counts X-rays received from a corresponding position in the object of interest *(such as positions 1 and 2 in Figure 9)*. Therefore the digital signals from the detector array give a density profile of the object's structure in the irradiated volume along the direction of the X-ray beam. The spatial resolution and signal sensitivity of the Slot Camera are determined by the various parameters discussed in the following sections.

1. SPATIAL RESOLUTION PARAMETERS.

The parameters for the Slot Camera geometry are shown in Figures 7, 8, and 9. These parameters are defined as follows:

r = length of the resolution element in the object of interest

s = width of the slot aperture

w = width of the detector element

L = length of the detector element

d = thickness of the detector element

a = width of the X-ray beam at the position of the slot aperture

b = thickness of the X-ray beam at the position of the slot aperture *(also designated as image slice thickness).*

$\boldsymbol{\theta}$ = angle between the intersection of the X-ray beam axis and the perpendicular to the plane of the slot aperture

1 Other scattering angles can be used according to the geometry used in the specific applications.

A = distance from the X-ray beam axis to the median plane of the slot aperture

B = distance from the entrance plane of the detector element to the median plane of the slot aperture

M = magnification ratio, B / A

V = $\mathbf{r^2 \cdot b}$ = Volume element *(voxel)* in test object which is source of scattered X-rays

D = distance from target of X-ray tube to midpoint of slot aperture

The spatial resolution, **R**, of the Slot Camera is determined by several of the above parameters. This resolution may be defined in terms of Line Pairs per cm, such that

$$\mathbf{R} = \mathbf{1/2r}$$

$$= \mathbf{M / 2s\,(1+M)} \qquad (1)$$

For a given value of **R**, the required slot width, s, is given as

$$\mathbf{s} = \mathbf{M / 2R\,(1+M)} \qquad (2)$$

Likewise for a given value of s or **R**, the required detector width, **w**, is given as

$$\mathbf{w} = \mathbf{s\,(1+M)}$$

$$= \mathbf{M / 2R} \qquad (3)$$

The preceding equations indicate that the resolution **R** depends primarily on the parameters, **M**, **s**, and **w**. Generally the size of the X-ray beam is determined by **r** such that the volume element *(voxel)* in the object is proportional to $\mathbf{r^3}$. For larger beam sizes such that the slice thickness, **b**, is greater than **r**, the value of **R** predicted by the above equations may be reduced.

The above relationships permit flexibility in the camera design such that the values of **M** and s may be adjusted to change either or both the camera size and resolution with tradeoffs that either reduce or increase the camera's signal sensitivity. This flexibility is a potentially important advantage of the Slot Camera over the conventional CT system to the extent that patient exposures and scan times can be greatly reduced.

2. SIGNAL SENSITIVITY PARAMETERS

The signal sensitivity of the Slot Camera may be defined as the image signal per pixel per unit X-ray exposure incident on the patient. The image signal per pixel is equal to the numbers of X-rays detected by a detector element in the Slot Camera as shown in Figure 9, where the detector elements, 1 and 2, receive the X-rays scattered from the corresponding volume elements at positions 1 and 2 in the object of interest. For a given detector and spatial resolution, the number of detected X-rays is linearly proportional to the quantity, $\eta\Omega$, where η is the X-ray detection efficiency of the detector element, and Ω is the solid angle subtended by the detector element from a corresponding point along the beam axis as shown in Figure 8 and 9, such that

$$\Omega \approx (wL)/(A+B)^2 \qquad (4)$$

According to the geometry used in the present measurements the value for η is equal to 0.0029. In the Phase II Program, the patient exposures required to produce a given image signal may be reduced by factors equal to the ratio $\eta_I \Omega_I / \eta_{II}\Omega_{II}$, where the subscripts I and II apply to the detectors and geometry used in the Phase I and Phase II programs respectively.

3. X-RAY DETECTORS

As indicated in Figure 9, the imaging panel consists of an array of twenty X-ray detectors which operate in a digital format such that each detector counts the X-rays incident on its entrance surface defined by the width, **w**, and the length, L. Each detector represents a pixel such that the spatial resolution is determined by the detector width, **w**, the slot width, **s**, and the resolution length, **r**, as discussed in Section A.1.

In this Phase I Program, each detector element is a plastic scintillator *(Bicron BC-400)*, such that **w**, **d**, and **L** are equal to 4 mm, 20 mm, and 15 cm respectively. Each scintillator is connected by a light pipe to a photo-multiplier tube as shown in Figure 10. The twenty X-ray detector elements were designed so that they are mounted adjacent to each other in a close-packed array as shown in Figures 11 (a), (b), and (c) where the adjacent detector assemblies do not interfere with each other. A front view of the detector array from the direction of the slot is shown in Figure 12.

As discussed in Section A.2, the detection efficiency, η , of each X-ray detector is an important parameter that determines both the minimum patient exposure and the scan time required to obtain a given spatial resolution. This efficiency, which is defined as the ratio of the number of X-rays counted by the detector to the number of X-rays incident on the entrance surface of the detector, was calibrated with a sodium iodide scintillation spectrometer. A 1 mm lucite scattering foil was placed at an angle of 45 degrees to the beam axis opposite the slot aperture in Figure 7, and the X-rays scattered at 90 degrees were counted first by the plastic detectors and then by the sodium iodide scintillation spectrometer *(3 inch diameter, 3 inch thick with a 1/2 inch aperture)* for which the X-ray detection efficiency was approximately unity. From these experimental results, the detection efficiency for the plastic detectors was determined to be approximately equal to 0.07.

With such a low value of 0.07 for the detection efficiency of the plastic scintillator, it is apparent that the sensitivity of the Slot Camera System can be greatly improved by using a detector with higher density and atomic number. For example, a comparison of the detection efficiencies of the above plastic scintillator and a sodium iodide scintillator *(2 mm wide, 8 mm thick, and 100 mm long)* is given in Figure 13 as a function of the discriminator pulse height setting which can be calibrated in terms of kilovolts. These efficiencies were estimated by carrying out statistical Monte Carlo calculations of the penetration and energy losses of mono-energetic X-ray in both the plastic and sodium iodide scintillators described above. In these measurements, most of the X-rays scattered at 90 degrees have energies in the region from 80 to 180 keV with a peak energy of approximately 110 keV. Accordingly as a first approximation, the Monte Carlo calculations were carried out for an X-ray energy of 100 keV. The results in Figure 13 show that the use of sodium iodide scintillators in place of the plastic scintillators can improve the detection efficiency, η , by an order of magnitude.

To obtain estimates of radiation cross talk between the closely spaced detectors in the array, additional Monte Carlo calculations were carried out for incident X-ray energies of 50 and 100 keV. The results showed that the counting rate for the sum of two detectors on either side of a radiation input detector is an order of magnitude smaller than the count rate for the irradiated detector. Also, the count rate for the sum of two detectors one set beyond the detector pair adjacent to the irradiated detector is an additional factor of two smaller than that of the adjacent pair of detectors. These results apply to both the plastic and sodium iodide detectors.

B. X-RAY SOURCE

In this Phase I program, the X-ray source was a 250 Kilovolt X-ray machine which was operated in the D.C. mode at 10 milliamps. The X-ray beam was filtered with a composite filter consisting of 5 mm aluminum plus 3 mm copper. The X-ray fluence spectrum for this filtered beam as measured in previous studies is shown in Figure 14. For some of the film measurements given in the results, only the 5 mm aluminum foil was used as a filter. Data on the spectral composition of the X-ray beam will be utilized in the future development of a prototype system, inasmuch as it may be desirable to modify the tube kilovoltage and filter composition in order to obtain an optimum spectrum which provides the smallest patient dose and scan times for the same diagnostic information in the X-ray image. A summary of the exposure measurements with this source will be given in the results.

In this Phase I program, the collimator for the X-ray beam was adjusted such that the beam dimensions, **a** and **b**, at position 1 opposite the slot aperture *(as shown in Figure 9 and defined in Section A.1)* were found from film measurements to be equal to 1.6 mm and 6.9 mm respectively. The dimensions of this beam can be adjusted for different applications in order to obtain the minimum scan times required for a given spatial resolution.

C. TEST OBJECTS

The test objects were designed to show the response of the Slot Camera to a periodic structure which can be interpreted in terms of the parameter, **R**, for spatial resolutions of 2, 3.3, and 5 Line Pairs per cm, which covers the region over which the conventional CT systems operate. The test object, which is designated here as a ***Spatial Resolution Indicator (SRI)***, has six lucite sheets each separated by an air gap equal to the thickness of the sheet, as shown in Figure 15. For spatial resolutions of 2.0, 3.3, and 5 Line Pairs per cm, the sheet thickness for the corresponding indicator is 2.5, 1.5, and 1.0 mm respectively. Each indicator is placed at an angle of 45 degrees to the incident X-ray beam and as shown in Figure 7, the X-rays scattered at approximately 90 degrees pass through the slot aperture to the detector array. Because of the limited number of detectors, the results will show that only 4 lucite sheets of the 2 Line Pairs per cm indicator could be imaged by the detector array.

IV. RESULTS

In accordance with the tasks described in Section III, a summary of the results is given in the following sections. These results provide experimental data for the spatial resolutions, the X-ray exposures and exposure rates, and the effects of body thickness. From these data, the minimum X-ray exposures and scan times are determined for the different spatial resolutions, and comparisons are made for the corresponding requirements in conventional CT systems.

A. SPATIAL RESOLUTION MEASUREMENTS

The experimental arrangement for carrying out the spatial resolution measurements is shown in Figure 7. It is important to note that no mechanical scan system is used in this arrangement because of the time and funding limitations in the Phase I program. Accordingly, data is obtained with the X-ray beam fixed at one entrance point to the test object, and the digital output of the detector array gives the density profile along the beam direction of the irradiated volume in the test object (without corrections for the beam attenuation in the object).

As a first test to demonstrate the potential imaging capabilities of the Slot Camera, a screen-film X-ray imaging system was substituted in place of the detector array shown in Figure 7, and a reproduction of the film images of the three Spatial Resolution Indicators *(SRI)* are shown in Figure 16 (a), (b), and (c) for 5, 3.3, and 2 Line Pairs per cm. These images were obtained with a 1 mm slot width and a 2 mm beam width for the 2 Line Pairs per cm SRI, and with a 0.5 mm slot width and a 1 mm beam width for the 3.3 Line Pairs per cm SRI. The raw images show the density profiles and the effect of the X-ray attenuation from the dark to the lighter bands as the beam passes through the SRI from the top to the bottom of the image.

In the second test, the digital outputs of the X-ray detectors are shown in Figure 17 for the 5, 3.3, and 2 Line Pairs per cm SRI test objects. The ordinate values are given in terms of the normalized counts per unit time which is equal to the ratio of the true count rate to a calibrated count rate in order to correct for the differences in the detection efficiencies of the detectors. These efficiency corrections were determined from detector response measurements with a standard X-ray source. The high and low values in these normalized count histograms closely parallel the high and low densities in the corresponding film images in Figure 16. The detector array used in this prototype system imposed certain restrictions on the digital output data shown in Figure 17. Because there are a limited number of detectors in the array, the length of the test object must be limited and for the Spatial Resolution Indicator corresponding to

2 Line Pairs per cm, only 4 instead of the 6 lucite sheets described in Section III C, were used. Also, to obtain the digital outputs for the Spatial Resolution Indicator corresponding to 5 Line Pairs per cm, it was necessary to reduce the detector widths from 4 to 2 mm by placing lead absorbers over half of the detector width and taking two separate exposures to cover the total image. Even with these reduced detector widths of 2 mm, there is an appreciable reduction in the modulation transfer function for the spatial resolution of the system, as shown by the reduction in the peak-to-valley count rations as the resolution increases from 2 to 5 Line Pairs per cm.

With an image processing system in the Phase II Program, these normalized counts will be corrected for the beam attenuation in the test object and will have the digital format necessary for image storage, image transmission, and video or hard copy image display. In addition, the software for the image processing system will provide the various image enhancement options such as contrast and edge enhancement, and/or histogram equalization. The final result will be an image display that accurately reproduces the structures appearing in any given region of the test object.

B. X-RAY EXPOSURE MEASUREMENTS

The X-ray exposure measurements were made for the following conditions:

a. The exposure rates were measured with a 6 cubic centimeter ionization chamber *(Model 10x5-180, MDH Industries Inc.)*, and are expressed in terms of the quantity, $\dot{\mathbf{X}}\mathbf{D}^2$, where $\dot{\mathbf{X}}$ is the exposure rate in Roentgens per second, and **D** is the distance in cm from the target of the X-ray tube to the midpoint of the slot aperture as shown in Figure 7.
b. The measurements were made with a 1.6 cm diameter collimator placed at the window of the X-ray tube, at a distance, **D**, equal to 100 cm on the beam axis.
c. The measurements were made for X-ray tube kilovoltages of 100, 200, 250, and 300 kilovolts in order to provide data that can be used to optimize the kilovoltage for a given application.
d. For each kilovoltage, measurements were made for two different filters placed before the collimator aperture. The first filter was a 3 mm aluminum foil, which was then replaced with a composite filter consisting of 3.2 mm copper plus 5 mm aluminum.

A summary of the values obtained for the quantity, $\dot{\mathbf{X}}\mathbf{D}^2$, for the different kilovoltages and filtrations is given in Figure 18. From these curves, the exposure rate can be determined for any given distance, **D**. Exposure measurements were also made with the highly collimated beam sizes of 1.6 x 6.9 mm that was used to obtain the digital output data in Figure 17. Because this smaller collimator transmits X-rays emitted from a fraction of the focal spot area, the exposure rates with the small collimator were found to be a factor of approximately 0.56 of the exposures values given in Figure 18.

C. BODY THICKNESS EFFECTS

As the X-ray beam penetrates to different body depths, there is an increase in multiple X-ray scattering processes and the general effect on image quality is to reduce the image signal to noise ratios *(or the image contrast)* in the portions of the image corresponding to increasing depths in the body. This general effect can be demonstrated with a lucite phantom having different gap separations as the penetration depth increases. This type of phantom was imaged with a screen-film system in place of the detector array which did not have enough detectors to image an object length of 20 cm as described in the previous section. The results are shown by the two images in Figure 19 which were obtained with and without a 7.6 cm lucite absorber interposed between the beam axis and the slot position. An inspection of these images which have not been enhanced by image processing techniques, leads to the following conclusions:

* The image signal or image contrast decreases as the penetration depth and absorber thickness increases.

* Spatial resolutions greater than 6 Line Pairs per cm are detectable at various body depths. Practical limits in determining the depths of body imaging may be controlled according to the kilovoltage of the X-ray machine and the size of the detector array.

* Improvements in image quality at body depths greater than 10 cm can be obtained with the use of anti-scatter baffles and with image processing techniques. A Phase II program will show how anti-scatter baffles can be designed and used in the Slot Camera to improve image quality.

* The effect of increases in body thickness is to increase the required X-ray exposures and scan times. Estimates of these increased values will be given in Section D.

D. MINIMUM DOSE REQUIREMENTS

The criterion for determining the minimum X-ray exposure, X_{min}, that is required to achieve a given spatial resolution is based on the attainment of a minimum value of ***five*** for the signal-to-noise ratio of the image signals obtained with the Spatial Resolution Indicators of 2, 3.3, and 5 Line Pairs per cm. The method of determining these minimum exposure values as a function of the spatial resolution is described in the following discussion.

The signal-to-noise ratio, σ, which is obtained with the exposure, X, measured at the position of intersection of the beam axis and the perpendicular extending through the midpoint of the slot aperture, is defined by the following equation:

$$\sigma(X) = \frac{n_1 - n_2}{(n_1 + n_2)^{1/2}} = (n_1)^{1/2} \frac{(1 - n_2/n_1)}{(1 + n_2/n_1)^{1/2}} \qquad (5)$$

where n_1 and n_2 are the number of detector counts at the peak and valley respectively of the histogram plots shown in Figures 17.

From the plot for a given spatial resolution, the ratio, n_2 / n_1, can be readily obtained by averaging the values for the peaks and valleys. Then the value of n_1 in Equation *(5)* is determined by converting the normalized counts shown in Figure 17, to the true count rate by using the calibrated correction factors for the efficiencies as discussed in Section III A.3. Accordingly the minimum X-ray exposure, X_{min}, required to produce a minimum signal-to-noise ratio, σ_{min}, of ***five*** is given by the equation:

$$X_{min}/X = \sigma_{min}/\sigma \quad \text{or} \quad X_{min} = 5X/\sigma \qquad (6)$$

From the experimental values obtained by the above procedure for n_1 and n_2, and the corresponding values for the exposure rates, $\dot{X}$, experimental values for the signal-to-noise ratios, σ, were obtained from Equation (5). Then the minimum exposure values required to obtain spatial resolutions of 2, 3.3, and 5 Line Pairs per cm were determined from Equation (6). With the conversion factor of one Roentgen equal to 0.88 rads for tissue, the minimum patient dose at the position on the beam axis opposite the slot aperture was determined for the above three spatial resolutions.

From the above results, the dependence of the minimum patient dose as a function of the Spatial Resolution is given in Figure 20.

These experimental curves were determined for a fixed geometry where Ω is equal to 0.0029 *(Section III A.2)*, a specific efficiency where η is equal to 0.07 *(Section III A.3)*, a fixed voxel size such that **V** is equal to 8 mm³, and for tissue absorption thicknesses of 10 cm and 20 cm for the scattered X-rays. These experimental curves are used as a basis of comparison for the results that can be obtained in a Phase II program with larger values for η and **V**. In fact, the minimum patient dose for a given spatial resolution can be reduced by the factor, **F**, where **F** is determined by the following equation:

$$\mathbf{F} = \frac{\Omega_{I} V_{I} \eta_{I}}{\Omega_{II} V_{II} \eta_{II}} \qquad (7)$$

where Ω , V, and η are defined in Section III A.1, with the subscript $_I$ indicating the values used in this Phase I program and the subscript $_{II}$ indicating the values that would be used in a Phase II program. Values of **F** for different spatial resolutions that can be obtained in a Phase II program are given in Table 1, and the minimum patient doses obtained with these **F** values are shown by the dotted lines in Figure 20 for tissue absorption thickness of 10 cm and 20 cm for the scattered X-rays. Compared to these curves in Figure 20, the average patient dose of 5 rads per slice that is required by the conventional CT system is shown by the dashed line.

Table 1. Dose Reduction Factor, **F**, for a Phase II Prototype

Spatial Resolution (Line Pairs per cm)	Program	Parameters η	Ω	V (mm³)	$\eta \Omega V$	F
2	Phase I	.07	.0029	8.0	0.0016	= 0.0087
	Phase II	1.0	.0030	63.0	0.19	
3.3	Phase I	.07	.0029	8.0	0.0016	= 0.038
	Phase II	1.0	.0019	23.0	0.043	
5.0	Phase I	.07	0.0029	8.0	0.0016	= 0.13
	Phase II	1.0	0.0013	10.0	0.013	

V. CONCLUSIONS

A. FEASIBILITY

The results obtained in this Phase I program have shown that with low efficiency X-ray detectors, the patient dose required by the Slot Camera to obtain spatial resolutions from 2 to 5 Line Pairs per cm, exceeds the average dose of 5 rads per image slice required by present CT systems. However with further prototype development in which high efficiency detectors are used *(sodium iodide for example)*, the Slot Camera can exceed the performance of present CT systems. The results in Figure 20 show that with high efficiency detectors, the local dose at the aperture position is 2.3 rads, and with exponential decay of the X-ray beam in the body, the average dose per slice is approximately 1.9 rads. Conversely if a high efficiency prototype is developed, the spatial resolution that can be achieved for the same average dose used by present CT systems, becomes greater than 7 Line Pairs per cm. Additionally, for oblique image slices or for partial slices, the average dose for the whole body may be at least an order of magnitude less than the present CT systems.

The above results apply to large body parts such as the chest or abdomen. For smaller body parts such as the head, neck, arms, or legs, the Slot Camera can greatly exceed the performance of the present CT system inasmuch as much smaller patient exposures are required for a given spatial resolution, or conversely much higher spatial resolution exceeding 10 Line Pairs per cm can be achieved for the same average body dose.

In summary, the above results establish the technical feasibility of the Slot Camera system. Compared to present CT systems, it offers low mass, low cube, low cost, low radiation burden for the body, and flexibility to permit tradeoffs in spatial resolutions, body doses, and scan times. In addition it will have the versatility to operate in conventional transmission or fluoroscopic modes, as well as the CT mode. These potential features of the Slot Camera system offer powerful advantages over present CT systems.

B. FUTURE PROTOTYPE DEVELOPMENT

As indicated in the preceding Sections, the Slot Camera System is ready for the development of a highly practical and useful prototype which can eventually be used for combat casualty care. This prototype will incorporate the following features:

* An image sensor array with high detection efficiency and with 100 detectors capable of imaging a 20 cm penetration depth.

* A small DC image processing system capable of performing various image enhancement processes as well as storing and networking archival images.

* A baffle system capable of greatly reducing multiple scattering effects for large body thickness.

* A mechanical Scan system that will provide simple partial or complete body scans.

* A dedicated X-ray source which will be incorporated into a scan head for the prototype system.

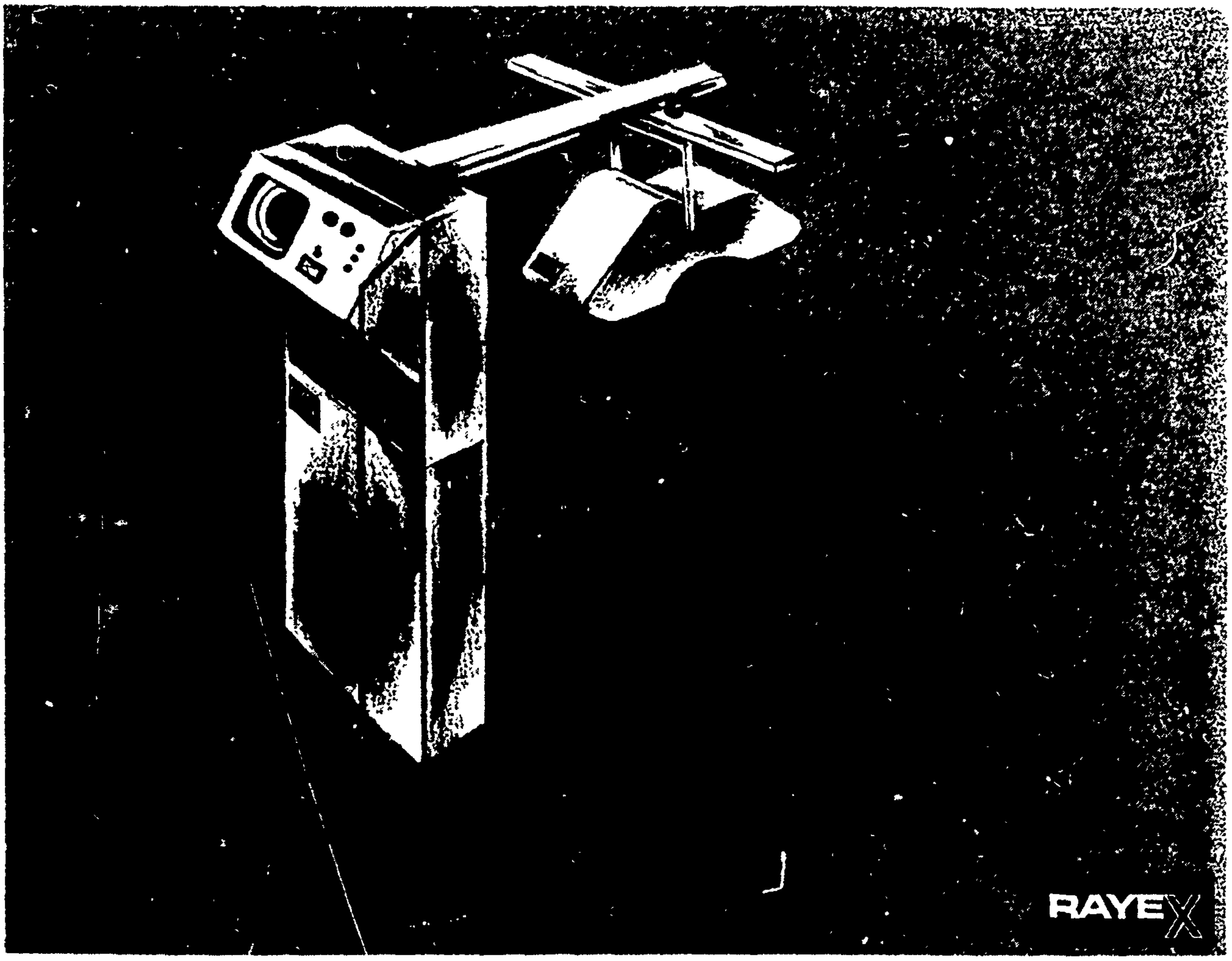


FIG. 1: Illustration showing the Rayex Multi-mode X-ray Imaging System positioned over a patient's chest ready to take complete or partial CT-type tomographic slices with arbitrary orientation, size, and shape.

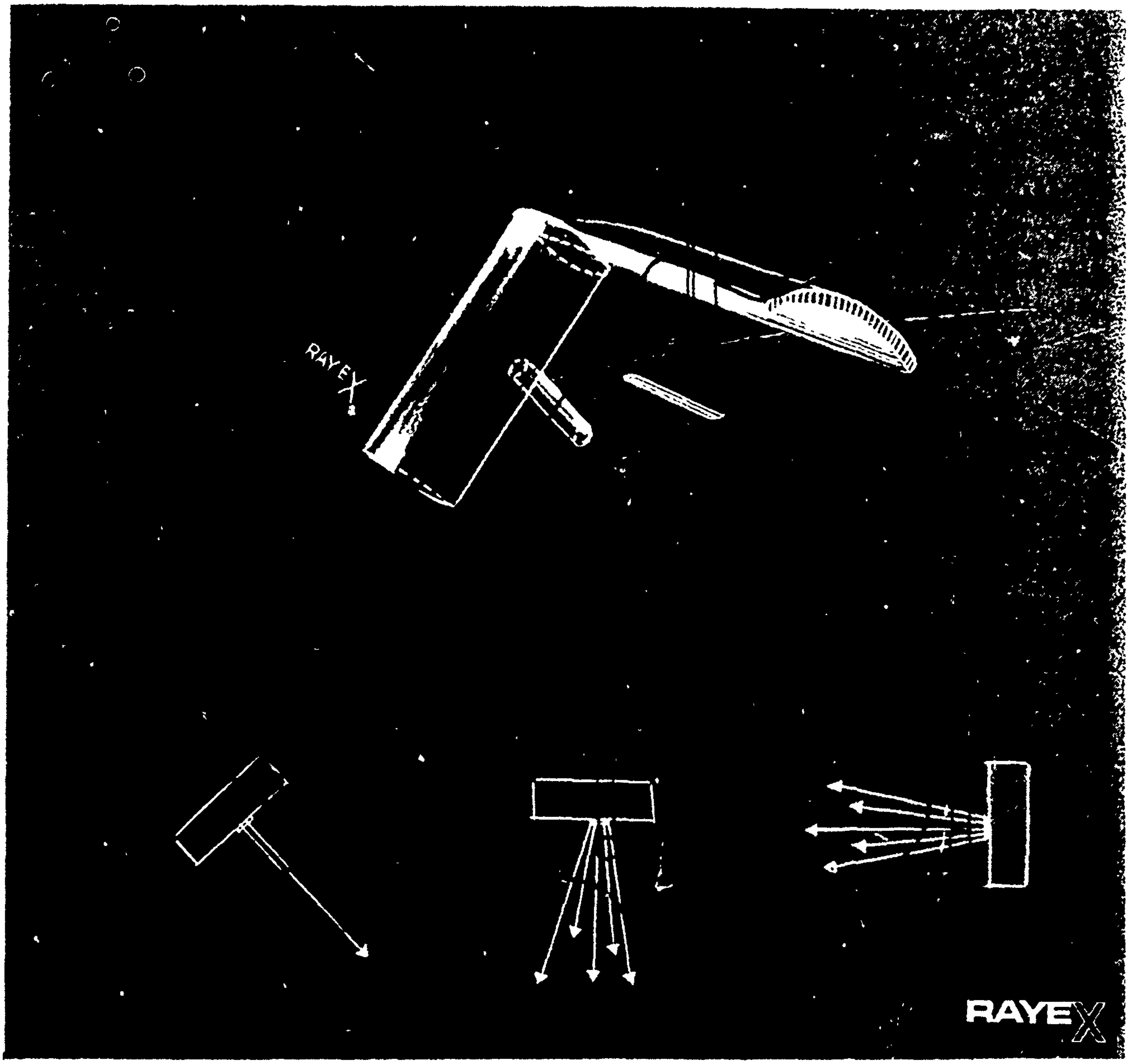


FIG. 2: Illustration showing the major components of the head assembly of the Rayex Multi-mode X-ray Imaging System. Sketches 1, 2, and 3 demonstrate representative positions of the fully adjustable head assembly as well as two types of x-ray beams which the head assembly can emit. Sketch 1 shows the pencil beam used for complete or partial CT-type tomographic slicing with arbitrary orientation, size, and shape. Sketches 2 and 3 show the conventional beam used for film radiographic flat-plane digital radiography, or real-time fluoroscopy.

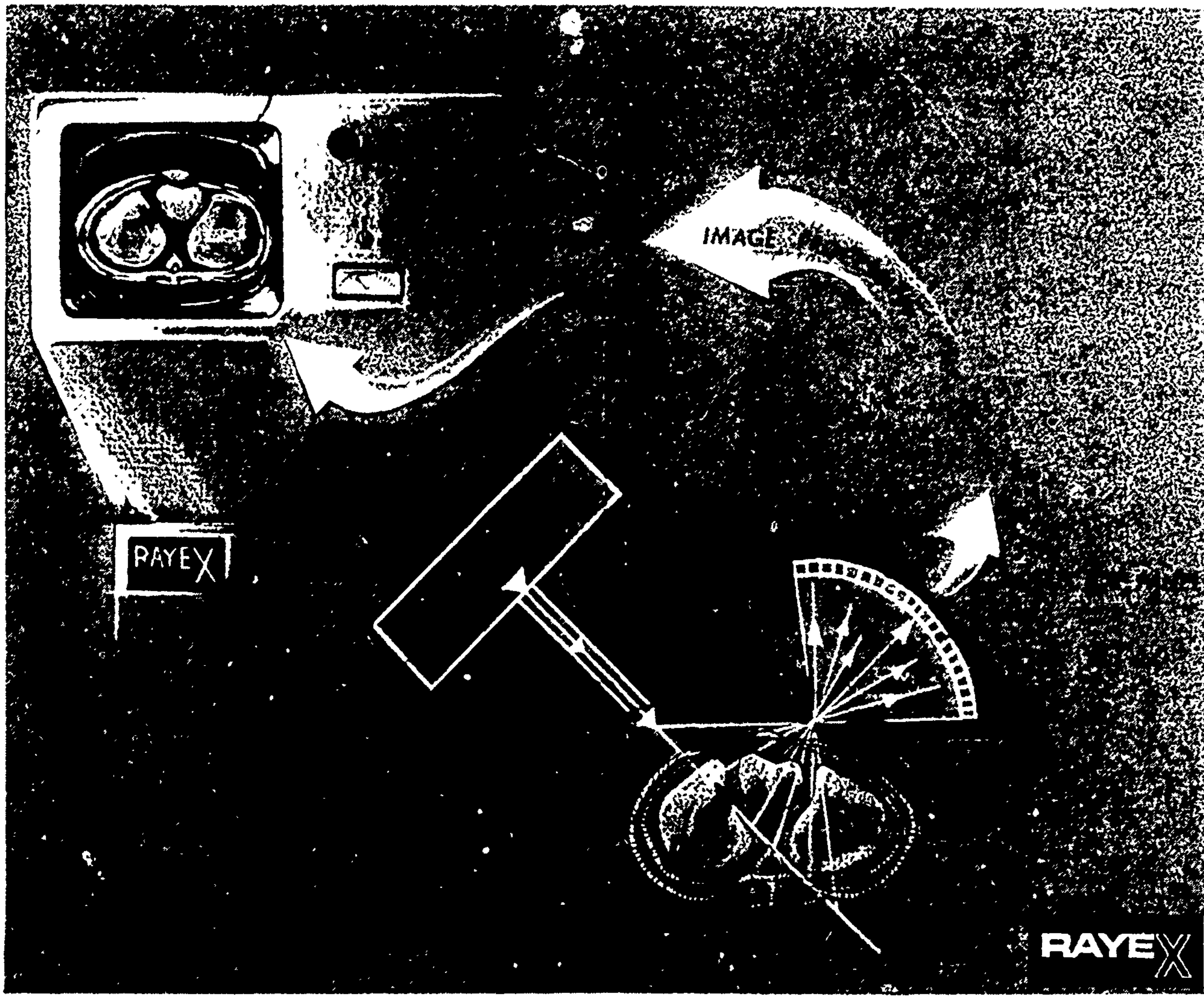


FIG. 3: Illustration demonstrating the operation of the Rayex Multi-mode X-ray Imaging System during CT-type tomographic slicing. The monitor shows the resultant CT-type tomographic image of a complete slice. The x-ray tube and collimator produce a pencil beam which passes through the patient's body. Backscattered radiation is produced at each point along the line of penetration. Some of this backscattered radiation enters the slot of the head assembly and is sensed by the elements of the detector panel as shown by the ray diagram. (Only those backscattered x-rays that pass through the slot are indicated in the figure). Each individual element of the detector panel selectively senses the x-rays which are backscattered from a given point along the pencil beam. The information sensed by the elements of the detector panel is then processed to produce the resultant image seen on the monitor. Note that the use of backscattered radiation permits one-sided operation and eliminates the need for an x-ray sensor on the back side of the patient. The black star-shaped object represents an embedded foreign body.

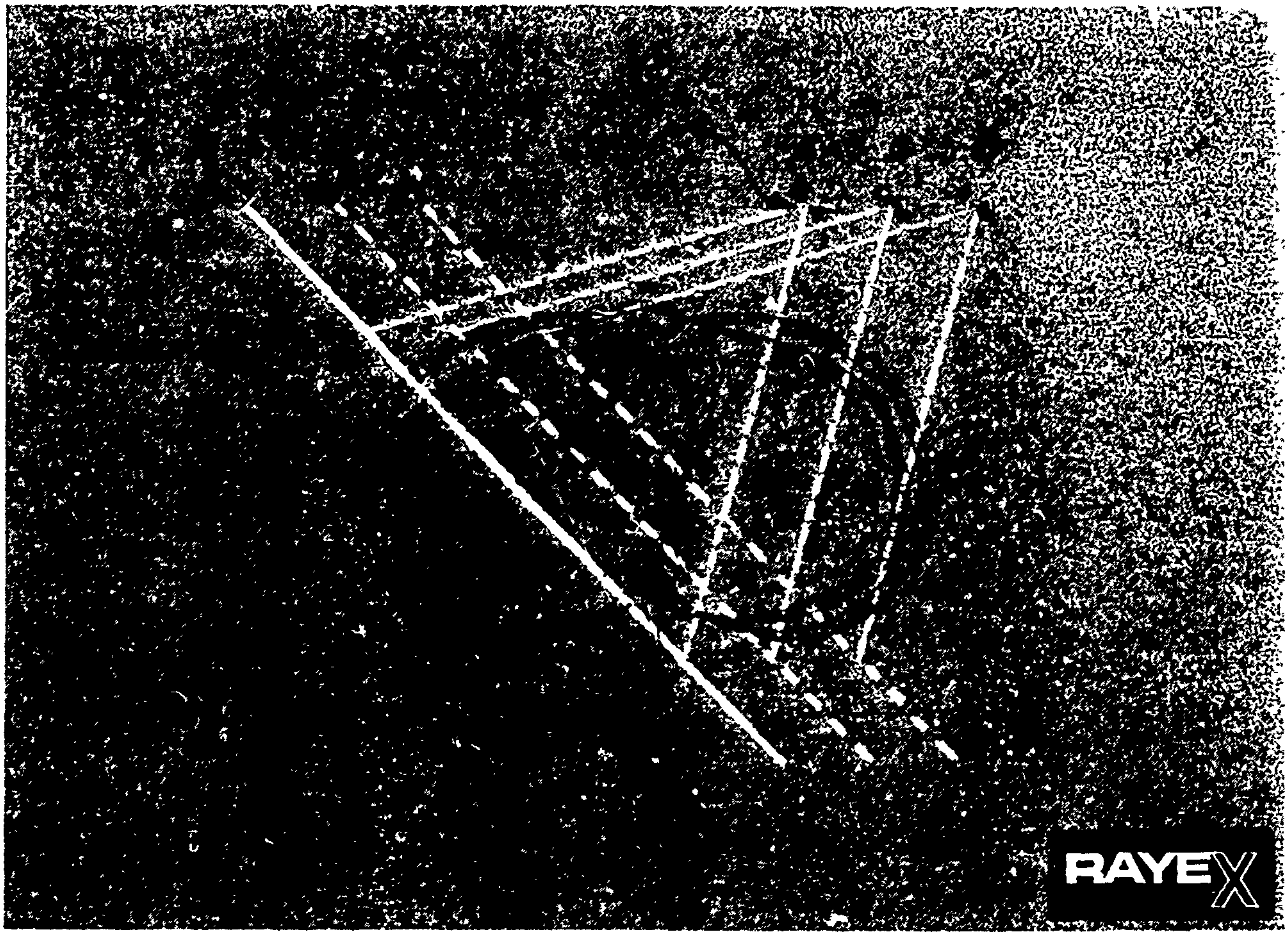


FIG. 4: Diagram showing the geometry relating the pencil beam, the backscattered radiation, and the slot at three representative positions, A, B, and C, during a typical scan. The black star-shaped object represents an embedded foreign body which is detected in position B. The Rayex Multi-mode X-ray Imaging System with its tomographic capability can readily detect non-metallic foreign bodies such as glass and plastic.

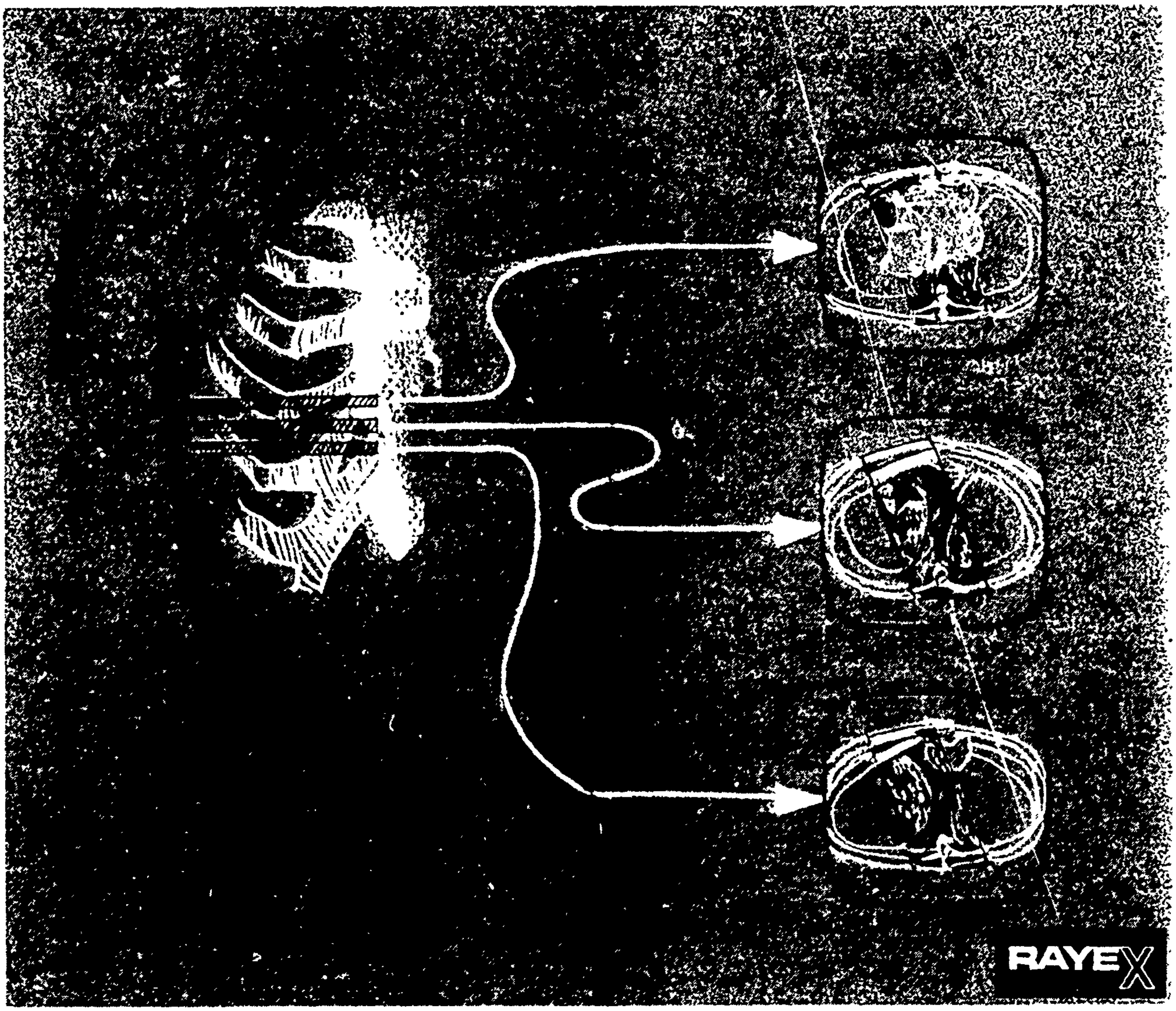


FIG. 5: Illustration demonstrating operation of the Rayex Multi-mode X-ray Imaging System for partial tomographic slicing. The partial slices (1, 2, and 3) are indicated on the torso and thier resulting images are shown on the monitors to the right. Note that the partial slices can be confined only to the critical regions of interest. The black star-shaped object represents an embedded foreign body.

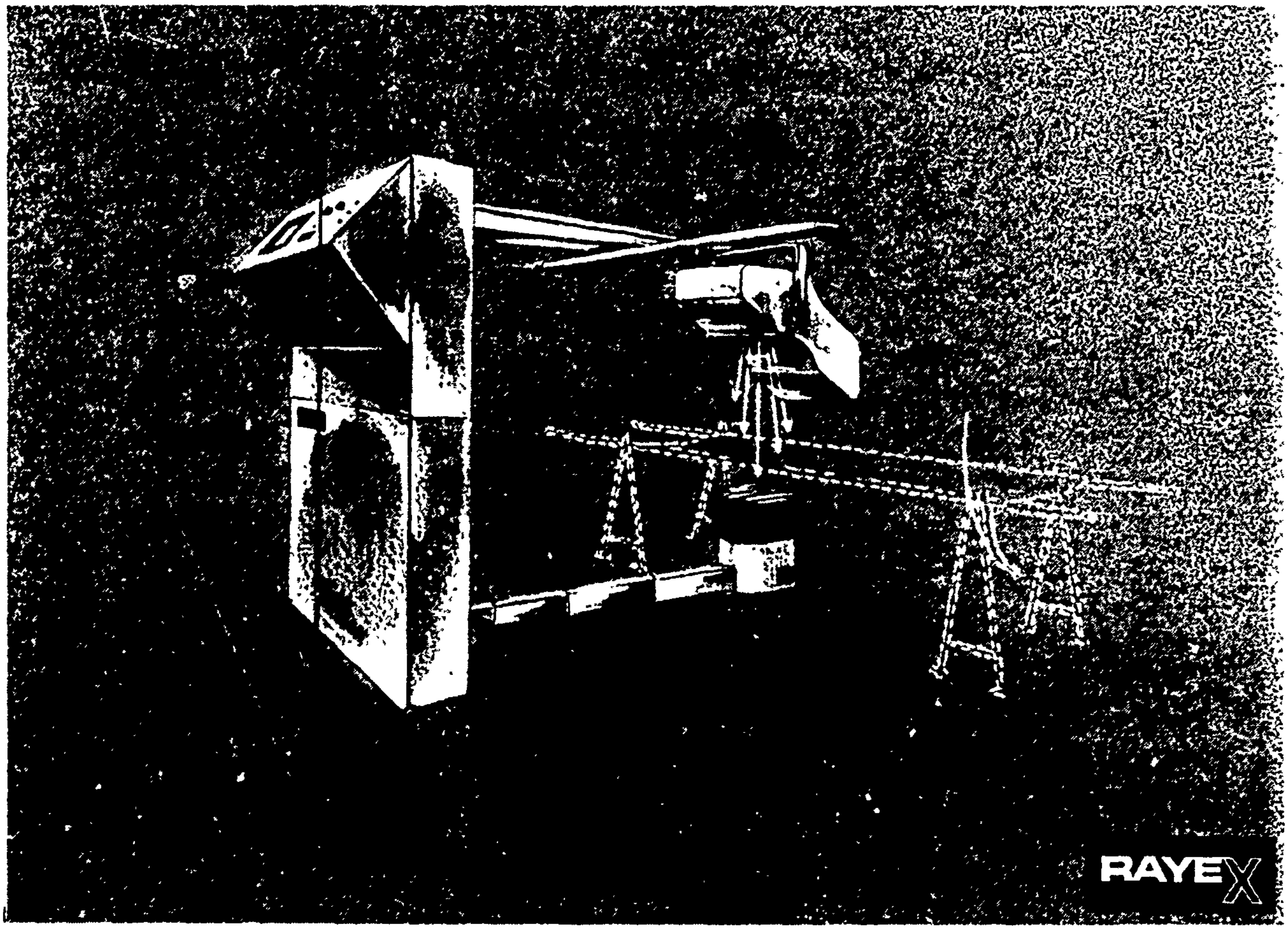


FIG. 6: Illustration showing the Rayex Multi-mode X-ray Imaging System positioned for real-time fluoroscopic imaging. Note that the head assembly is emitting a conventional beam. Also note the mobile image intensifier below the litter. The same configuration can be used to produce a flat-plane digital image. The image intensifier can also be replaced by film cassettes for conventional film radiography.

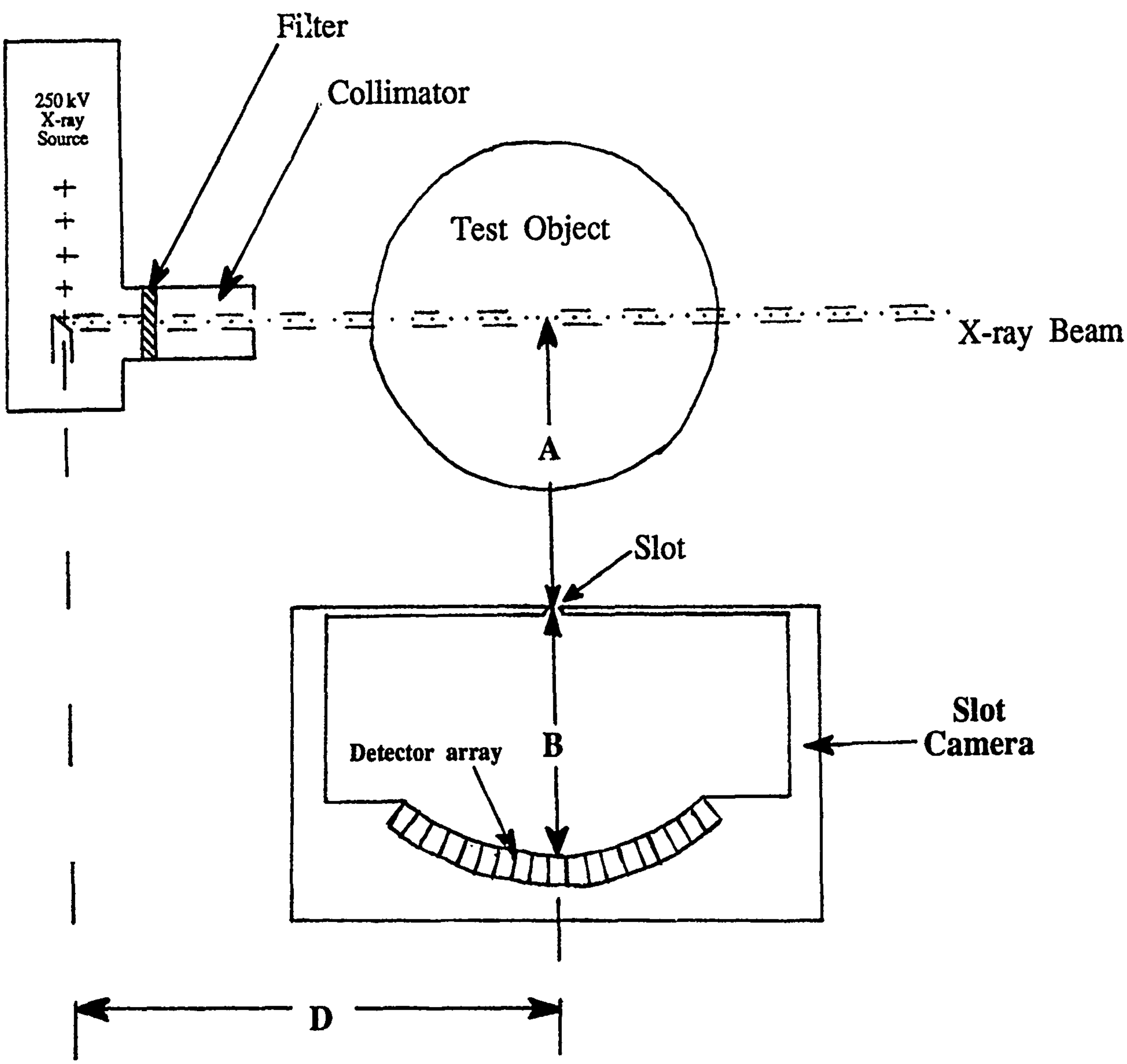


Figure 7. Experimental Arrangement for the Slot Camera System

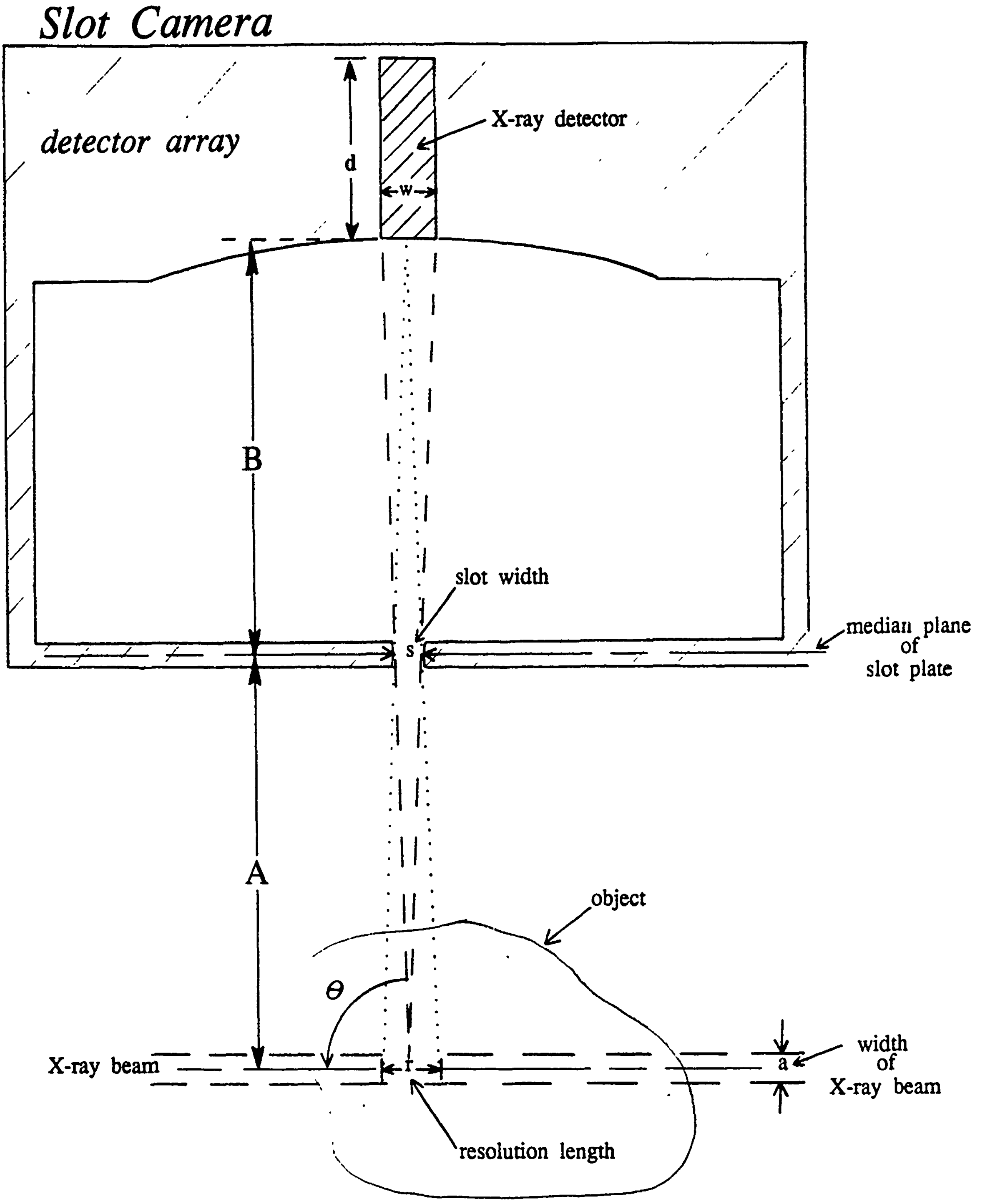


Figure 8. Parameters for Slot Camera Geometry

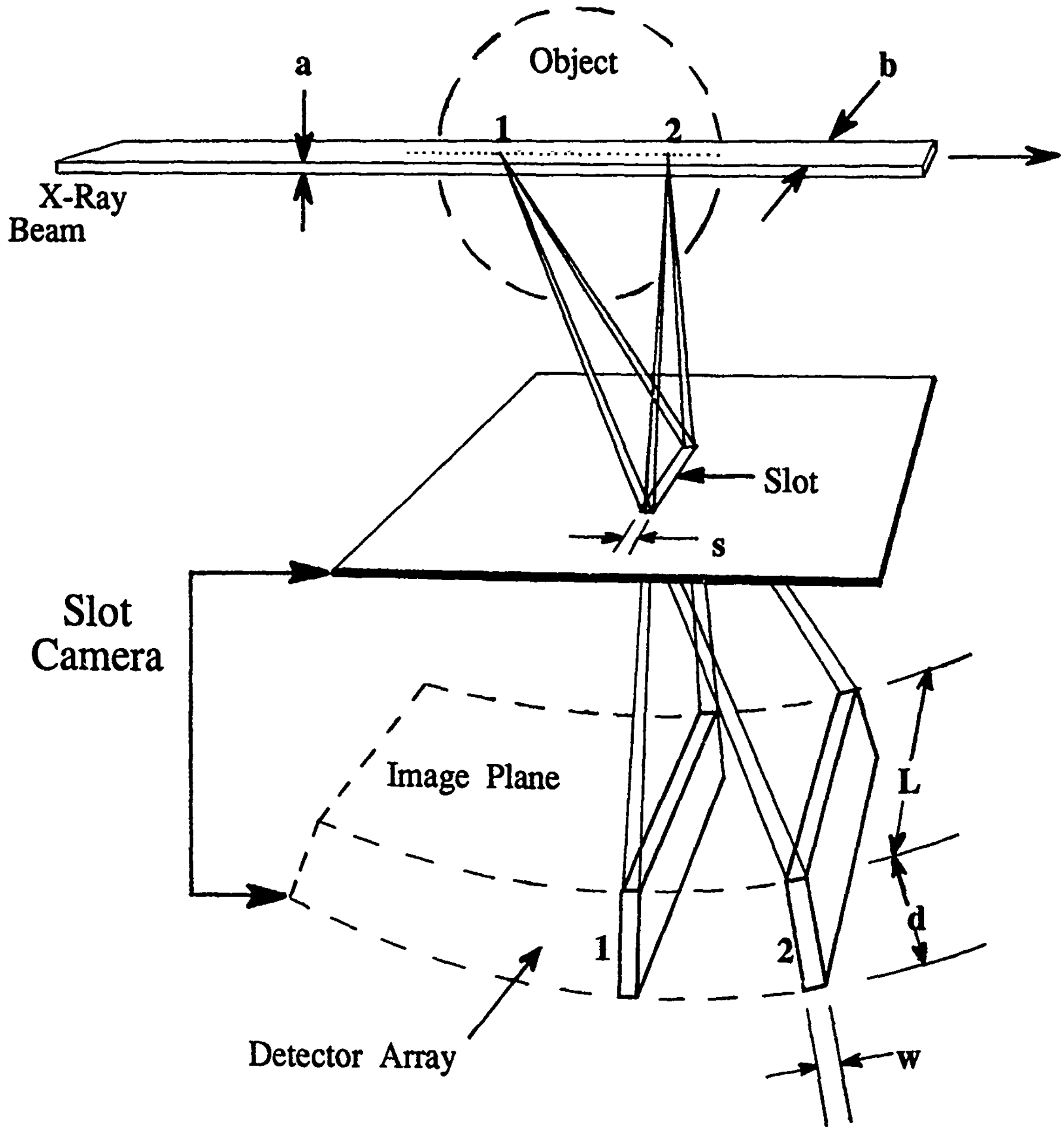


Figure 9. Detector geometry for Slot Camera. Detectors 1 and 2 are equivalent to pixels in the image plane corresponding to the volume elements at positions 1 and 2 in the object.

Figure 10. X-ray detector consisting of plastic detector, light pipe, and photo-multiplier tube.

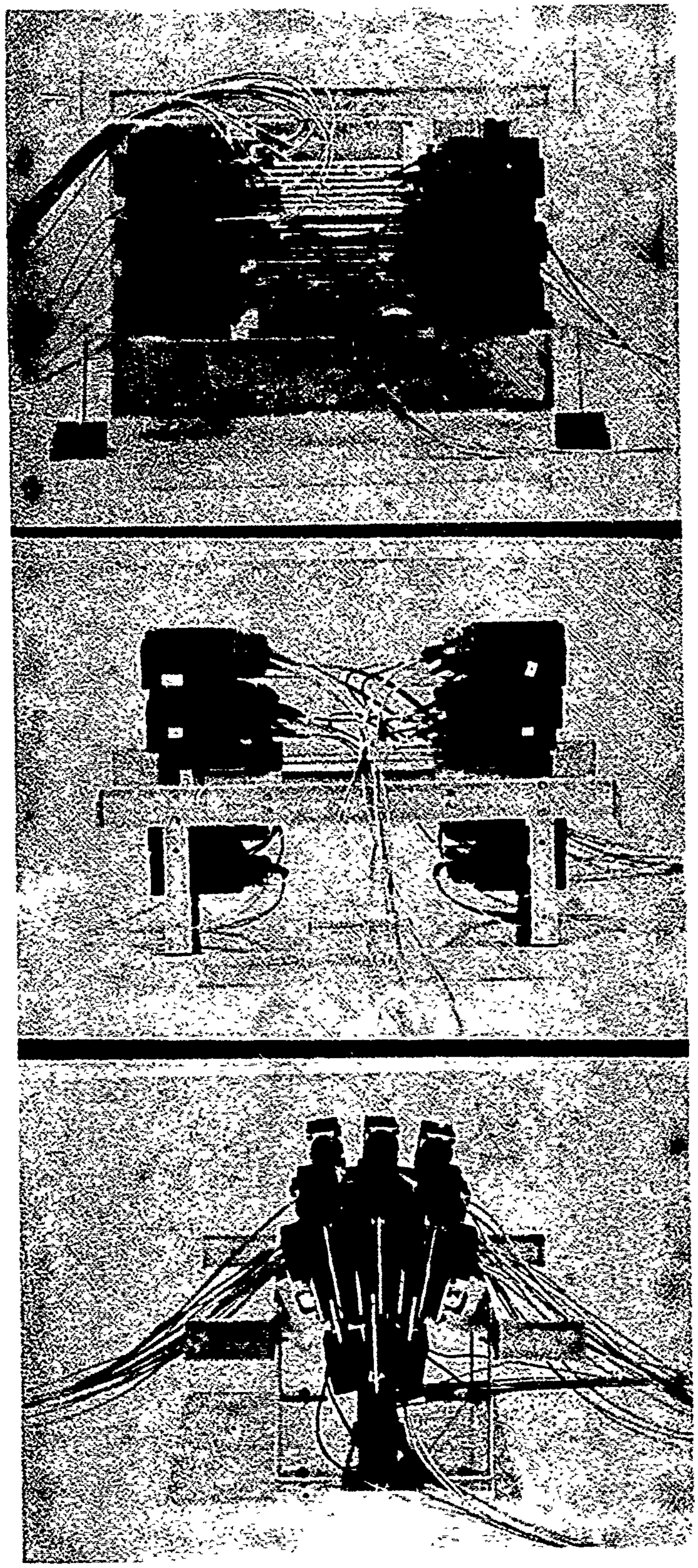


Figure 11. Assembly of 20 detector elements with different views to show packing arrangement.

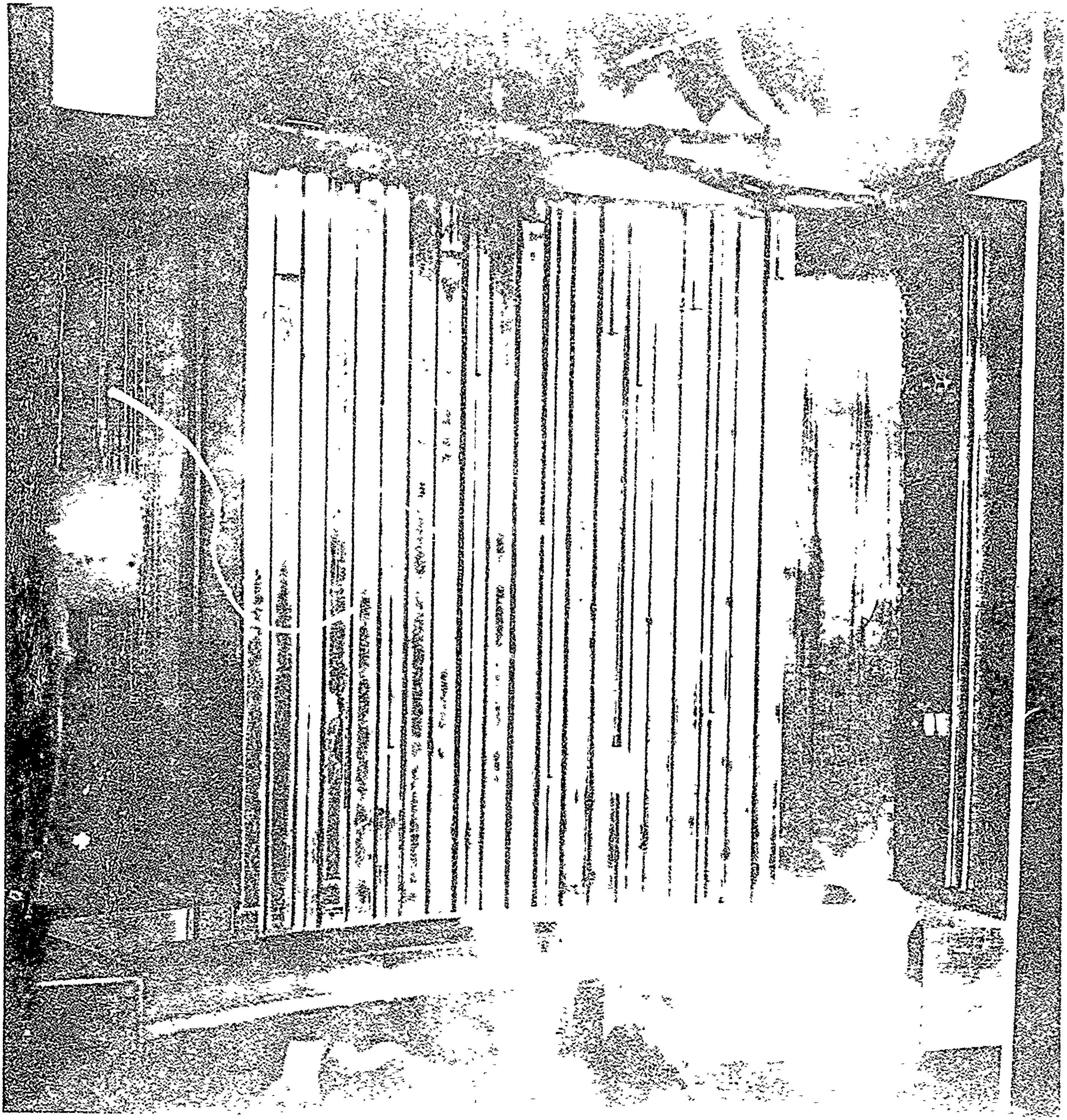


Figure 12. Front View of the twenty [illegible] (*front slot direction*). Each [illegible] long and 4 mm wide

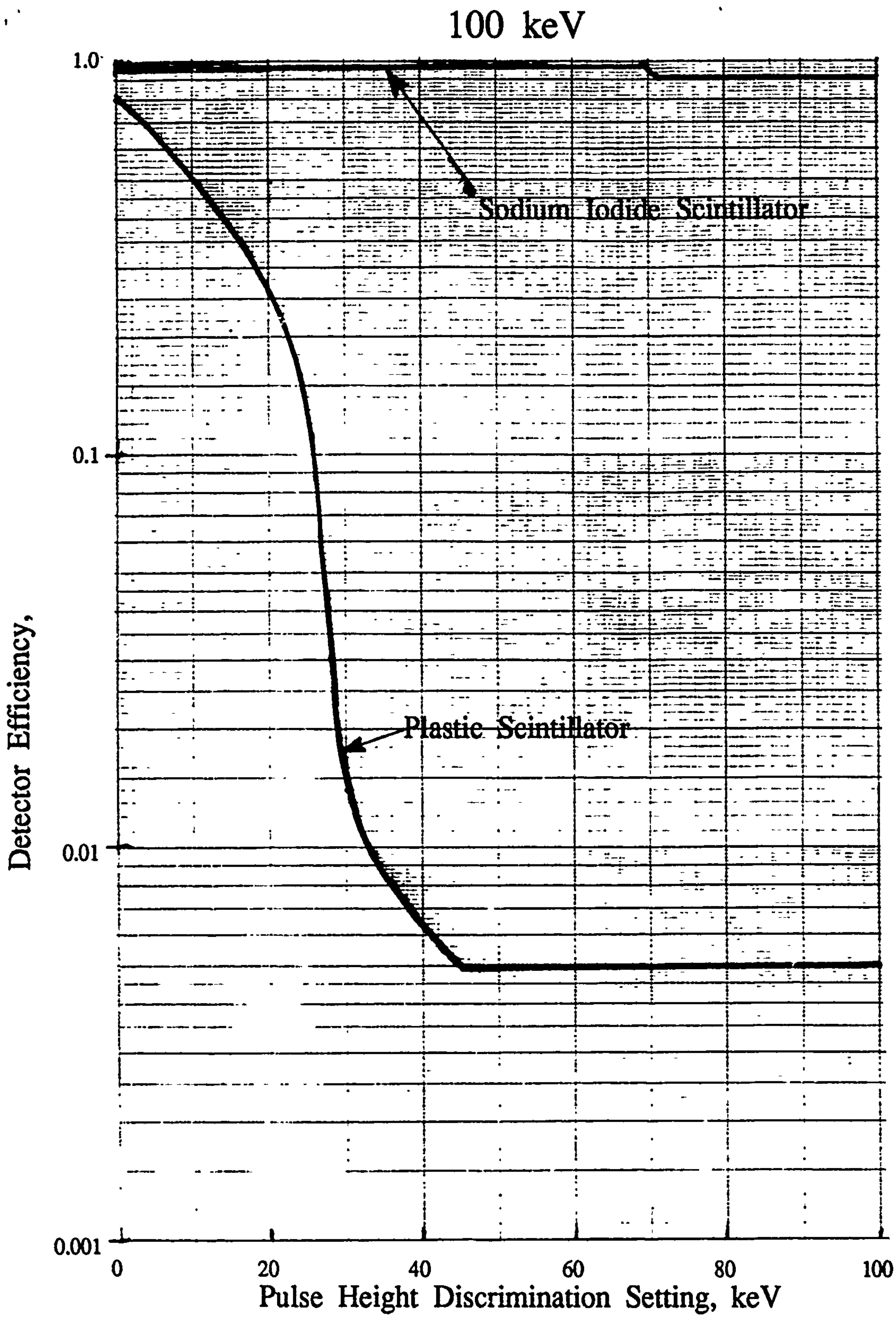
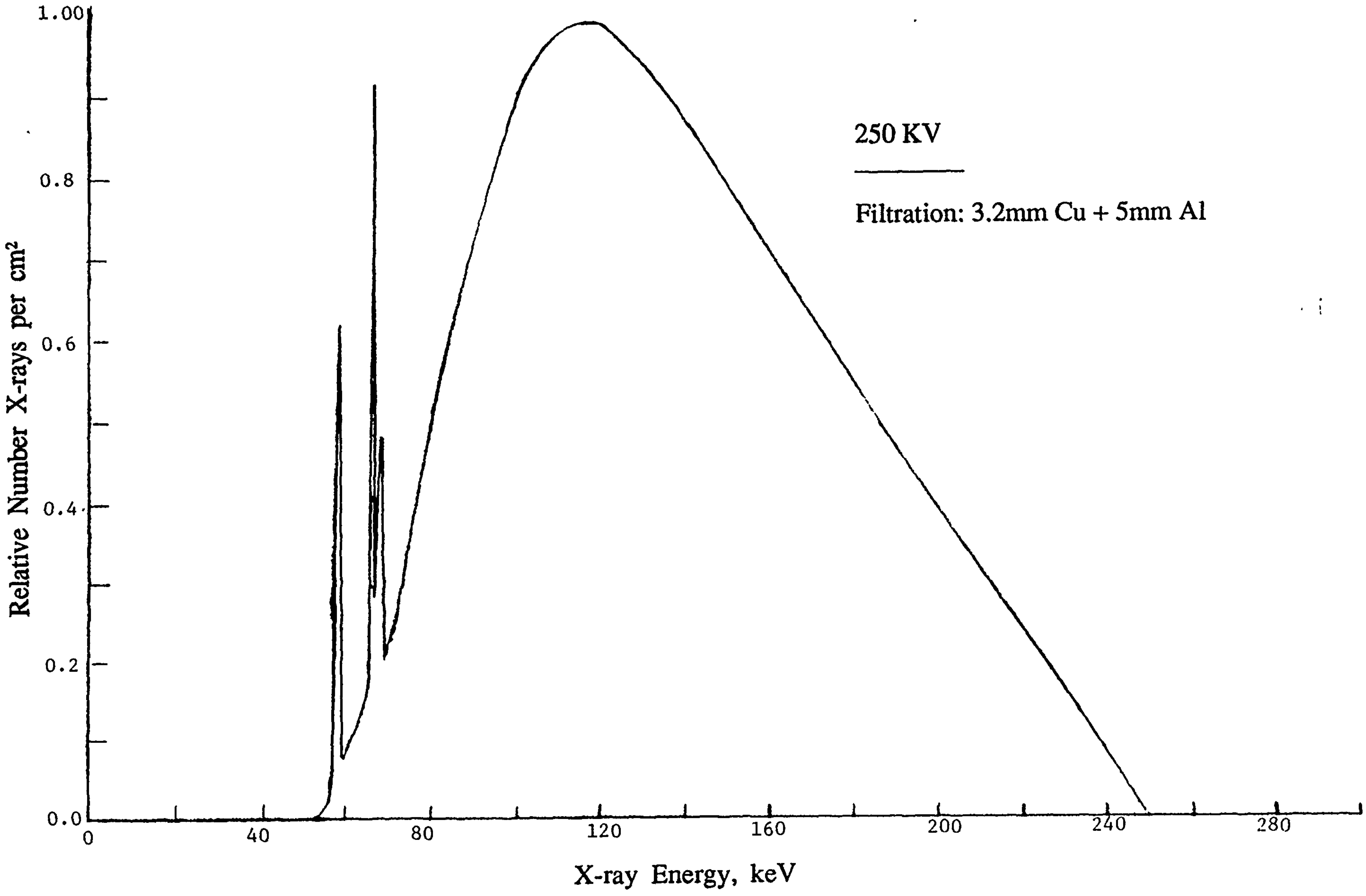


Figure 13. Dependence of the X-ray detection efficiency on the minimum pulse height accepted by the counting circuit.

Figure 14. Fluence spectrum for X-ray beam produced by 250 kV X-ray machine with 3.2 mm copper plus 0.5 mm aluminum filtration.



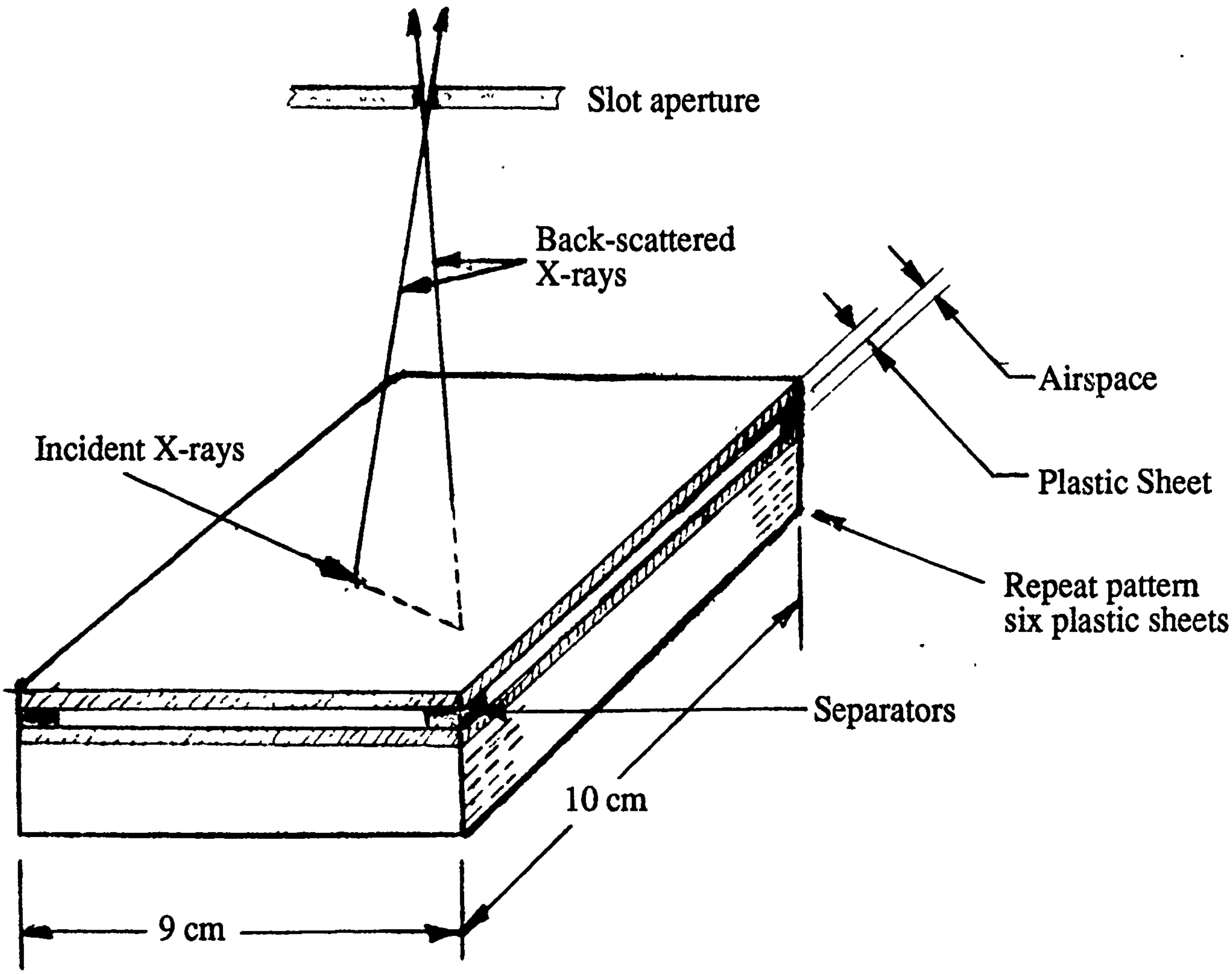
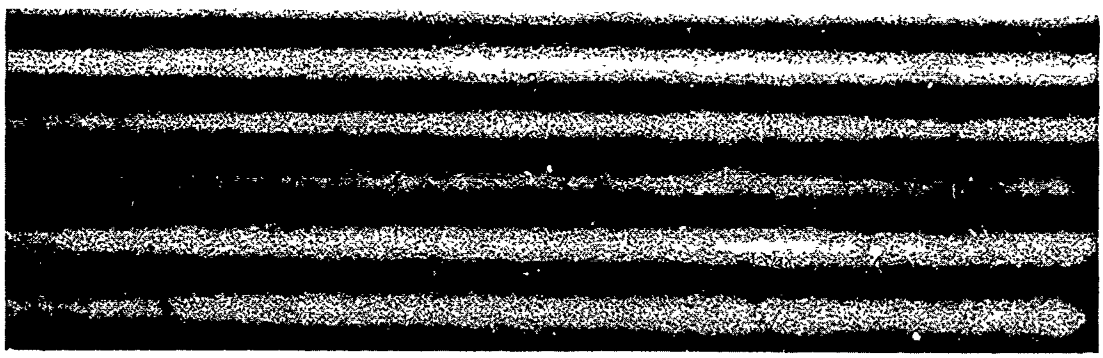


Figure 15. Structure and arrangement of the Spatial Resolution Indicator.

5 Line Pairs per cm

3.3 Line Pairs per cm



2 Line Pairs per cm

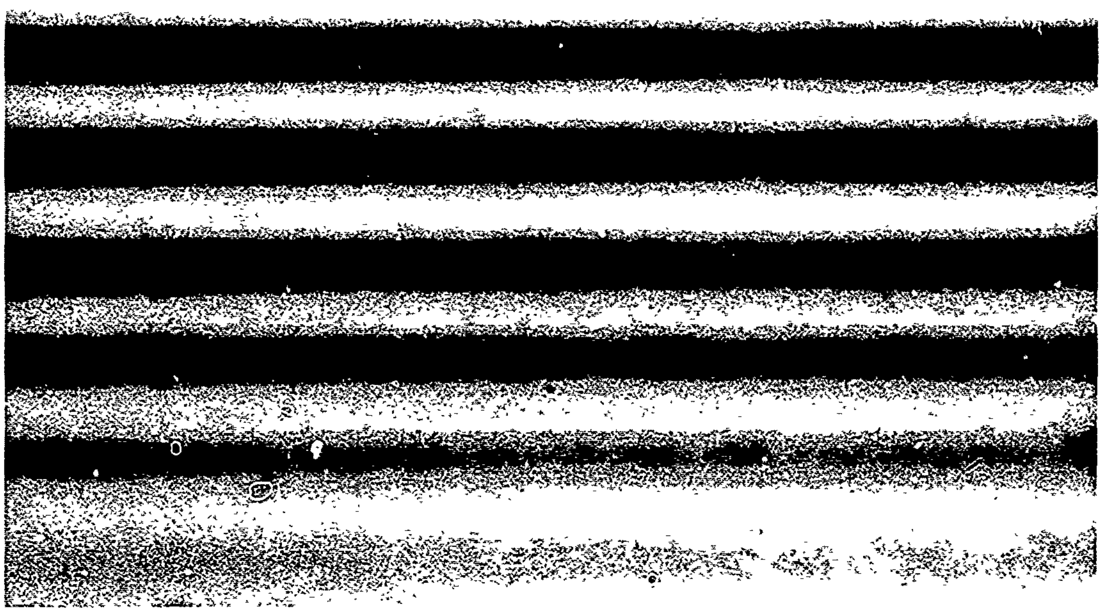


Figure 16. Reproduction of film images obtained with the Slot Camera. The test object is the Spatial Resolution Indicator for 5, 3.3, and 2 Line Pairs per cm.

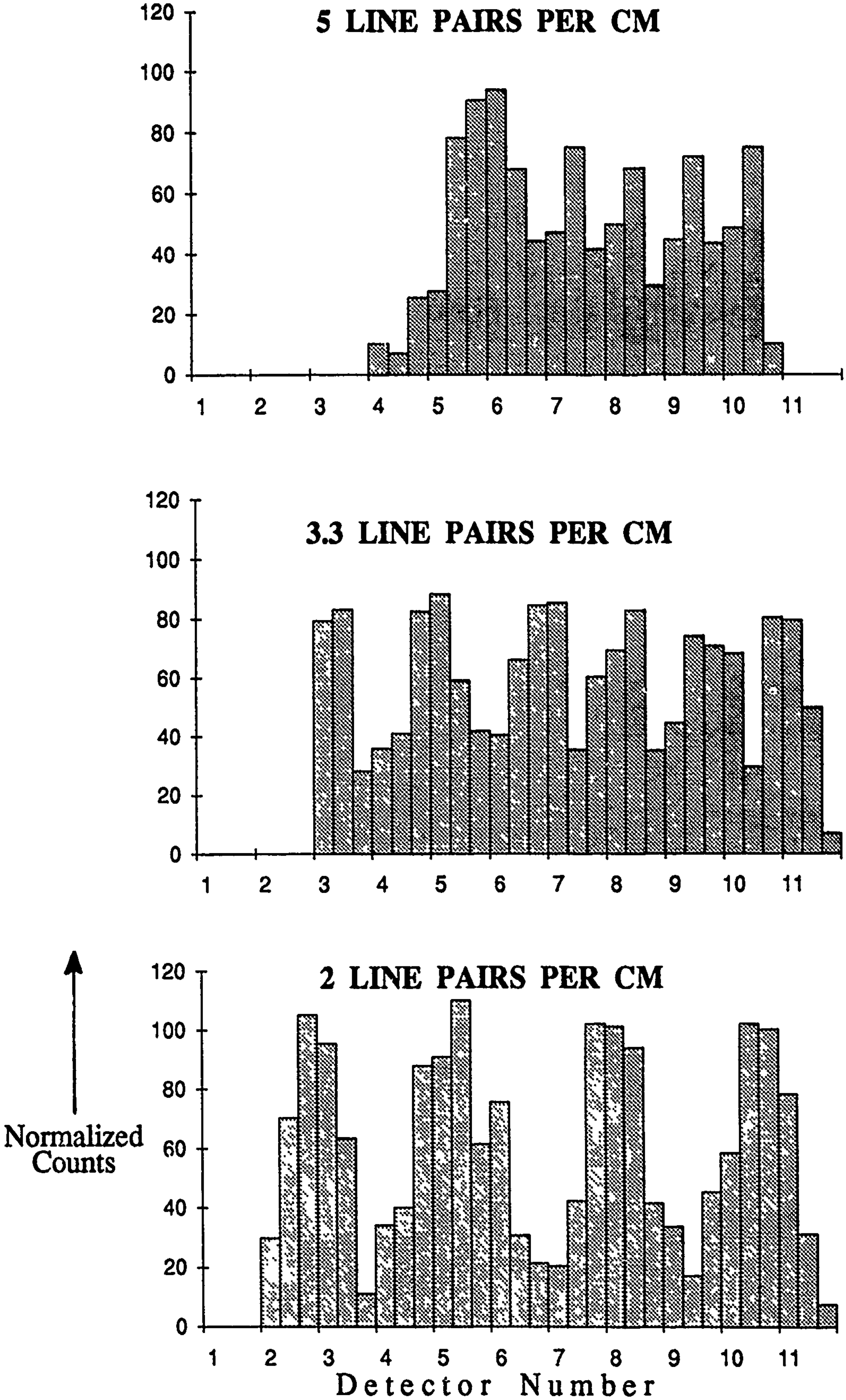


Figure 17. Digital outputs from the Slot Camera detector array with the Spatial Resolution Indicators corresponding to 2, 3.3, and 5 Line Pairs per cm.

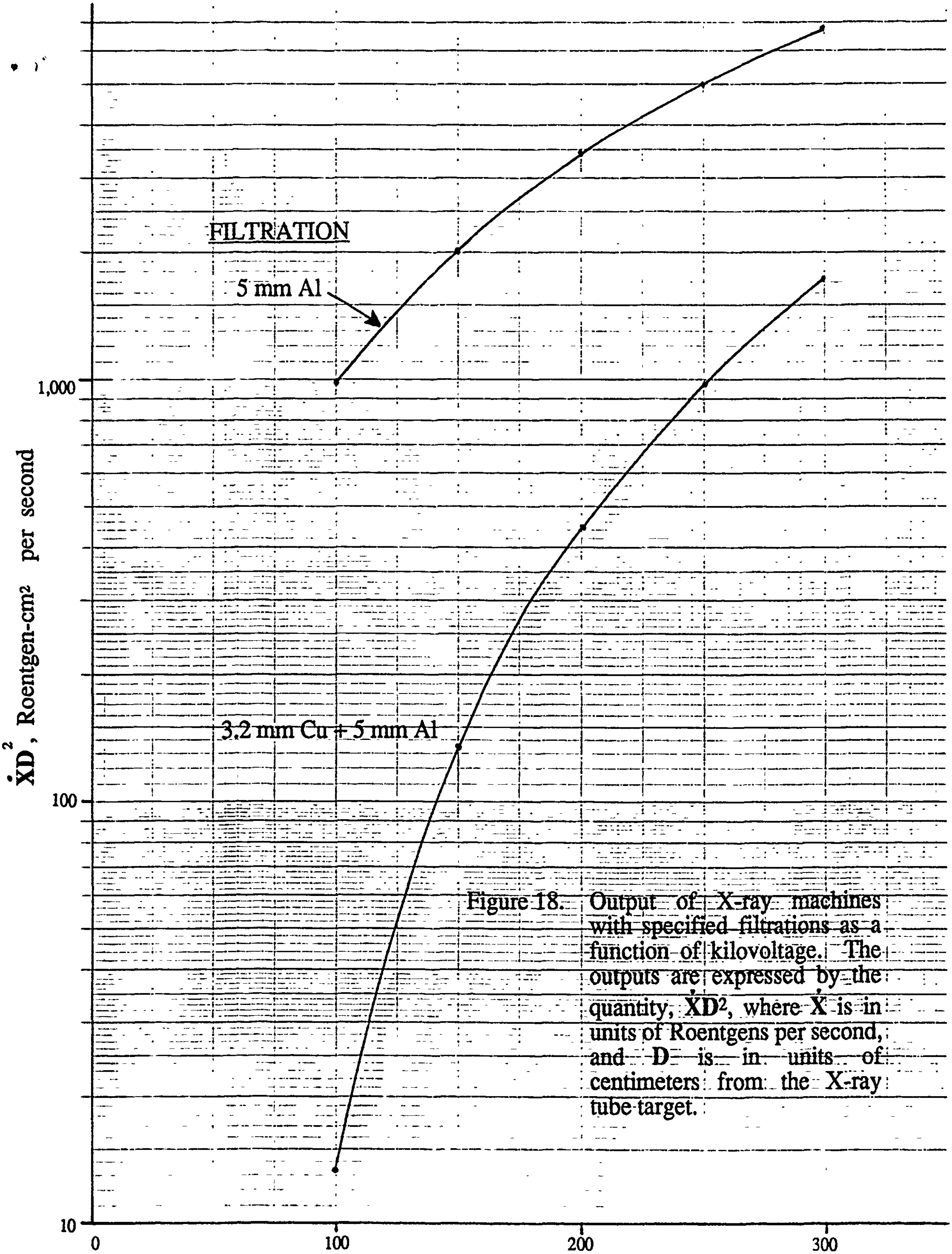


Figure 18. Output of X-ray machines with specified filtrations as a function of kilovoltage. The outputs are expressed by the quantity, $\dot{X}D^2$, where $\dot{X}$ is in units of Roentgens per second, and D is in units of centimeters from the X-ray tube target.

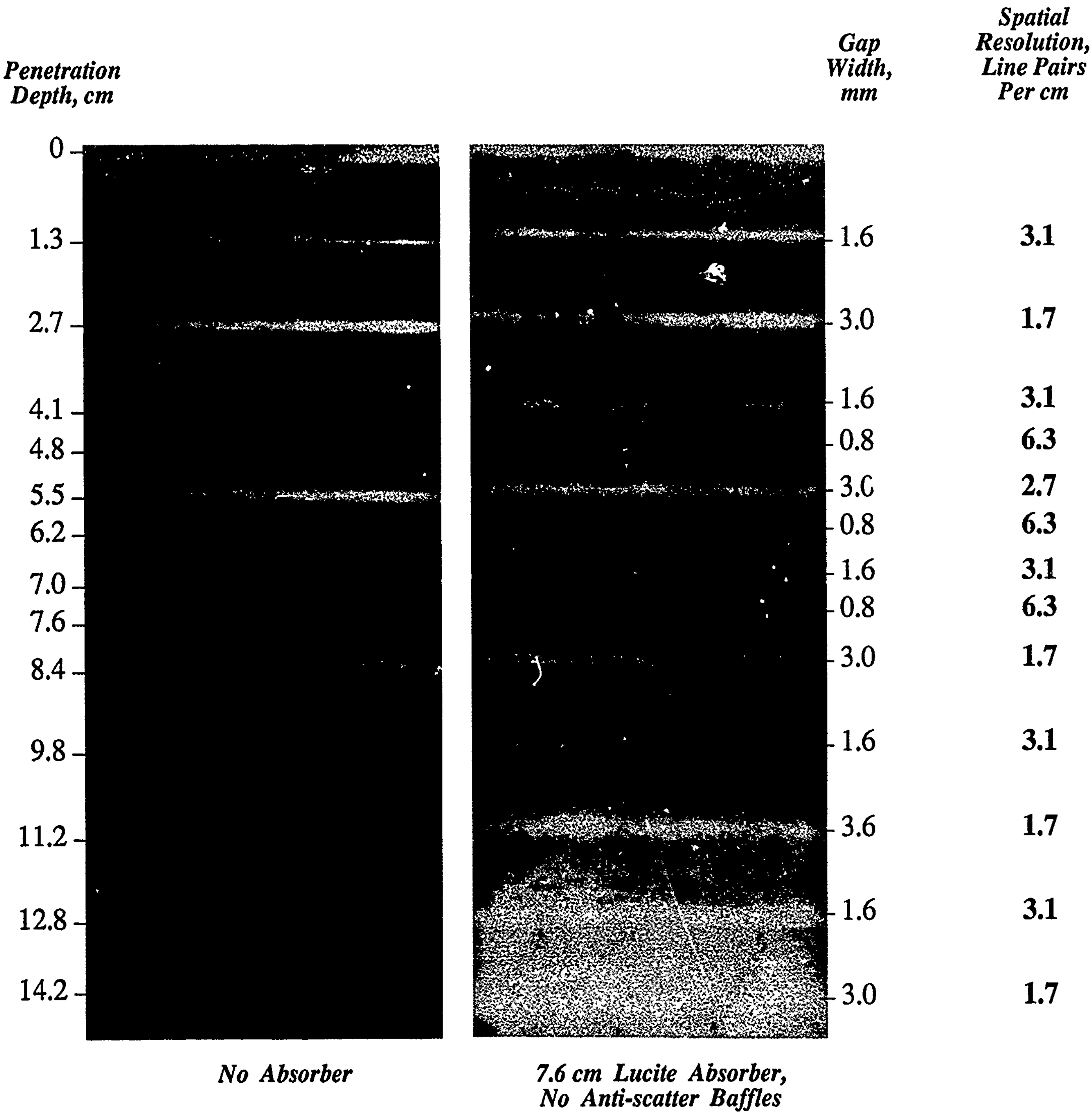


Figure 19. Reproduction of film images of a 15 cm long lucite test object with different gap widths at different depths. Images are compared with and without a 7.6 cm lucite absorber interposed between the beam axis and the slot camera.

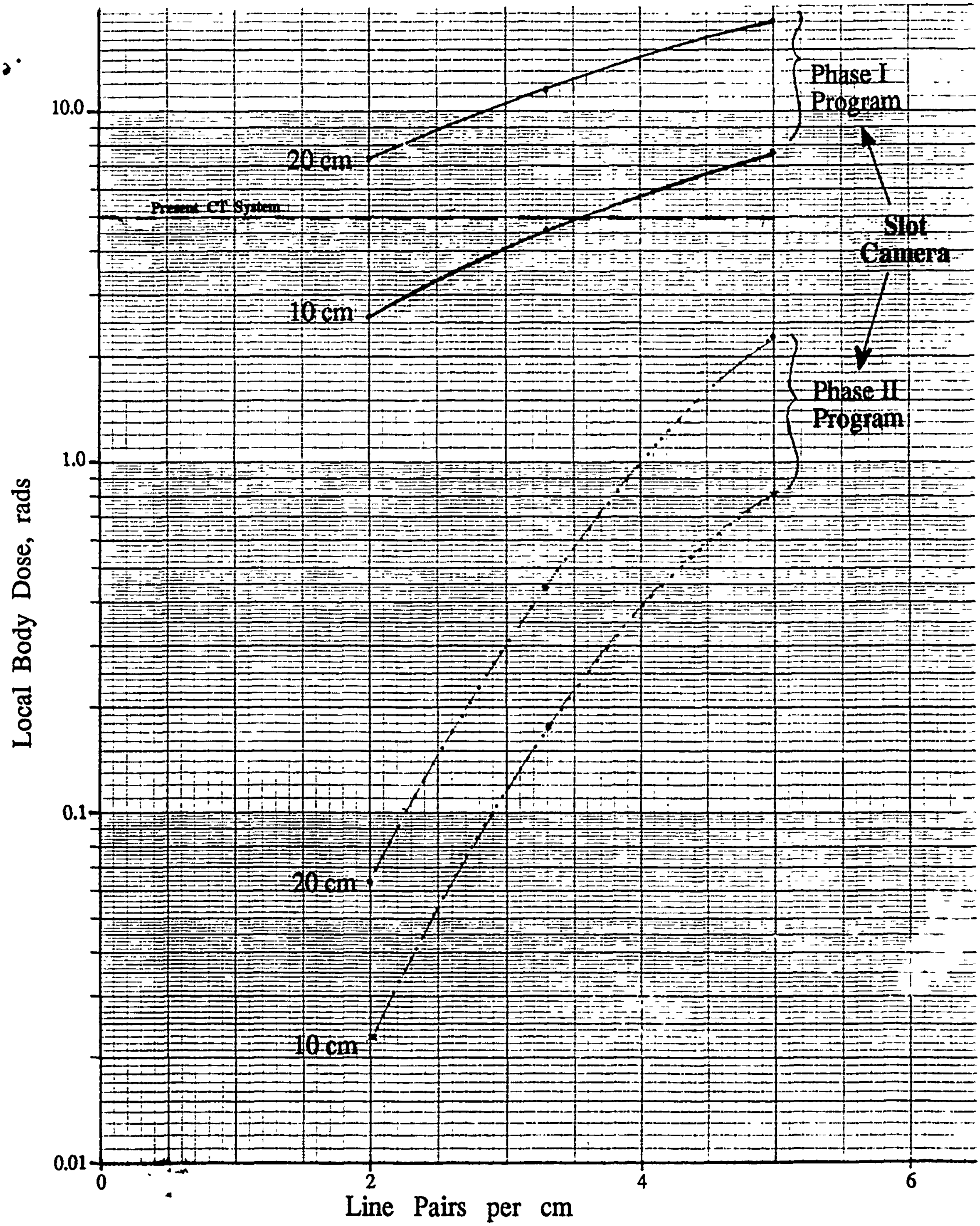


Figure 20. Local body dose at position of beam axis opposite slot aperture *(Figure 7)*, as a function of spatial resolution *(Line Pairs per cm)*, for X-ray body absorption thickness of 10 and 20 cm. The solid lines were obtained with the low efficiency detectors (η = *0.007)* used in the Phase I program. The dotted lines were obtained with high efficiency detectors (η = *1.0)* which are proposed for a Phase II program. The dashed line is the average dose per image slice for present CT systems.